Lasers in Surgery and Medicine 23:7–17 (1998)

Thermic Effects in the ''Vestibule'' During Laser Stapedotomy With Pulsed Laser Systems

S. Jovanovic,[1]* U. Schönfeld,[1] R. Fischer,[1] M. Döring,[1] V. Prapavat,[2] G. Müller,[2] H. Scherer[1]

[1]ENT Department, Benjamin Franklin Medical Center, Free University of Berlin, D-12200 Berlin, Germany
[2]Institute for Medical/Technical Physics and Laser Medicine, Benjamin Franklin Medical Center, Free University of Berlin, D-12200 Berlin, Germany

Background and Objective: **Apart from the ablation properties at the stapes footplate, the degree of thermic loading in the inner ear is important in determining the suitability of pulsed lasers for stapedotomy. The aim of the study is to compare the thermic effects in the vestibule with different pulsed laser systems.**
Study Design/Materials and Methods: **Temperature increases and heat exchange processes in the fluid (physiological saline) were examined in a calorically and physiologically approximated cochlea model for applying the laser parameters effective in creating footplate perforations.**
Results: **With all systems, increases in the energy density, number of pulses, and thus resultant total energy lead to higher temperatures. In the effective energy density range, the highest temperature increases achieved with the requisite number of pulses at a distance of 2 mm behind the perforation are 26°C with the Ho:YAG laser. The lowest temperature maxima are 5.5°C with the Er:YAG and <5°C with the Er:YSGG laser. The excimer laser, investigated at only one energy density, showed maximal temperatures of 10°C.**
Conclusion: **The Er:YSGG and Er:YAG laser can be applied in laser stapedotomy in a relatively broad energy density range without a risk of inner ear damage by thermic loading. On the other hand, the Ho:YAG laser is not recommended for stapedotomy because of the higher energy density and pulse rate required for sufficient perforation and the resultant higher temperature increases in the perilymph. Though likewise achieving perforations with only slight temperature increases in the fluid of the cochlea model, the excimer laser does not seem appropriate for stapedotomy because of the long period of heat exposure (ca. 60 s) due to the lower ablation rate at the stapes necessitating a longer application time. Lasers Surg. Med. 23:7–17, 1998.**

Key words: **excimer laser; Ho:YAG laser; erbium lasers; cochlea model; temperature measurements; inner ear**

Contract grant sponsor: German Research Foundation; Contract grant number: JO 180/1-1.

*Correspondence to: Dr. Sergije Jovanovic, Freie Universität Berlin, Universitätsklinikum Benjamin Franklin, Hals-Nasen-Ohrenklinik mit Polikliniken, Hindenburgdamm 30, D-12200 Berlin, Germany. E-mail: schoenf@zedat.fu-berlin.de

Accepted 13 April 1998

INTRODUCTION

Stapedotomy with the laser has now become an established method for surgical management of otosclerosis. The thermically acting argon, KTP 532, and CO_2 lasers [8,12,17,22] in continuous-wave (cw) mode and the CO_2 laser in superpulse mode [15] are currently in clinical use for laser stapedotomy. Low laser powers, pulse durations, and pulse repetition rates are selected, since possible noxious effects cannot be excluded when higher doses are applied [11].

More recent experimental studies demonstrate that, apart from the continuous-wave lasers, some pulsed laser systems (holmium (Ho)YAG, erbium (Er)-YAG, and Er:YSGG laser) are also suitable for stapedotomy [13].

Compared to continuous-wave lasers, pulsed lasers result in photoablation at impulse durations in the nano- and microsecond range, which reduces the total energy needed for stapes footplate perforation and thus also lowers the thermic tissue exposure [7,13,14,18,21,23]. Thermic loading of the inner ear, however, is based not only on the initial temperature developing at the footplate but also essentially on the heat exchange processes resulting in the cochlea (perilymph) [4,11].

The aim of this study is to compare the pulsed laser systems relevant for stapedotomy with respect to possible heating of cochlear structures when perforating the footplate under comparable and reproducible measuring conditions. The parameters were selected so as to achieve an adequately large perforation of the human footplate.

MATERIALS AND METHODS

Temperature measurements were done in the model rather than in a human petrous-bone preparation because it afforded better reproducibility and thus quantitative comparability of the results with various laser systems and also enabled direct observation of the mechanisms (Fig. 1a). In accordance with the heat parameters of the cochlea (bone: $\lambda = 0.2$ W/mK; $c_w = 1{,}300$ J/kgK) [2], the model consists of acrylic glass ($\lambda = 0.184$ W/mK; $c_w = 1{,}440$ J/kgK) with a cylindrical volume of 0.1 ml (Ø = 3 mm, 1 = 14 mm) filled with physiological saline kept at 37°C. The temperature increases were measured perpendicularly at distances of 1 mm, 2 mm, and 3 mm behind the perforation with a NiCr-Ni thermocouple (2ABAC 025 TM, Philips Co.) of low heat capacity and rise time (Ø = 250 μm, $t_r < 10$ ms) (Fig. 1b).

The stapes footplate was replaced by a 90-μm-thick compact-bone comparable to the stapes with respect to the laser-beam absorption properties (Fig. 2) and the perforation effect (perforating diameter, shape and structure of the perforation, reproducibility of the perforation diameter, and thermically altered marginal zones) [13]. Measurement of the transmission properties of a human stapes in the UV (300–400 nm) and in the middle IR (2.5–22 μm) wavelength range shows no absorption bands in the UV range and a marked transmission reduction in the IR range at 3 μm, 5 μm, and 10 μm (with a minimum below 10 μm). A similar transmission spectrum is also found in the compact substance of a human femur with an absorption maximum within the IR range at a wavelength of 10 μm (Fig. 2).

Use of a high-speed videocamera (Ekta Pro Motion, Kodak Co.) yielding up to 1,000 images/s permitted the additional documentation of heat-transport mechanisms indirectly based on induced air or gas bubbles transported by the convection flow.

Four different laser systems were used (Table 1): excimer laser (MAX 10LP, Technolas Co.), holmium:YAG laser (Spektrum Co.), erbium:YSGG laser (Spektrum Co.), and erbium:YAG laser (MBB Co.).

The laser beam was applied via a micromanipulator or directly.

The settings shown in Table 1 were selected according to the laser parameters we had previously determined in vitro to be optimal for achieving a sufficiently large perforation (diameter: 0.5–0.6 mm) of the human stapes footplate. To avoid applications of higher total energies, however, the energy densities must be chosen in such a way that the requisite perforation can be achieved with the smallest possible pulse count. We therefore define as effective the lowest energy density range achieving adequate perforation diameters [13].

Five measurements were recorded at each setting, and the median value and scattering range were determined.

RESULTS

Perforation of the footplate with pulsed lasers results in strong warming of the perilymph near the footplate. Use of a high-speed videocamera yielding 1,000 images/s demonstrates the re-

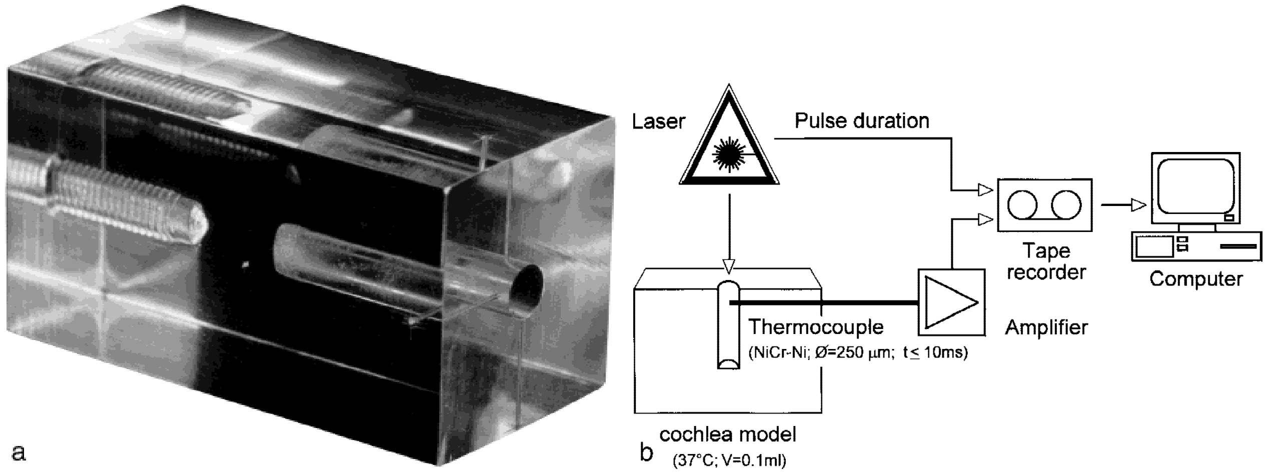

Fig. 1. **a:** Cochlea model. **b:** Test arrangement for temperature measurements.

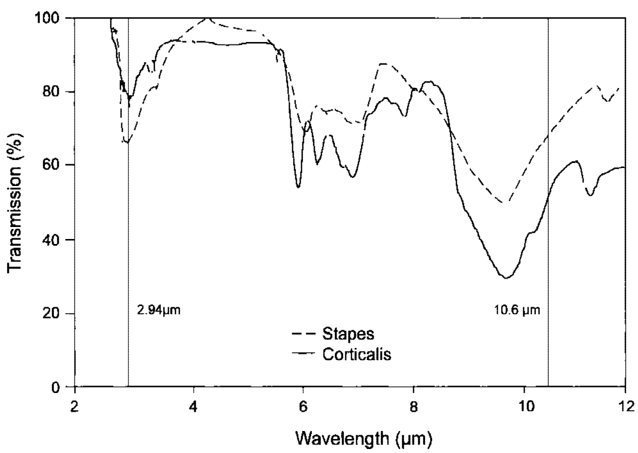

Fig. 2. Transmission spectra of human stapes and cortical substance in the infrared range.

sultant heat-exchange processes indirectly based on induced gas bubbles. This is exemplified by a single Er:YSGG laser impulse passing through the perforated bone platelet directly into the fluid (Fig. 3). Already 1 ms after the laser shot, the explosive photoablative effect of this pulsed laser system generates a short convection flow axially directed at a high velocity of up to about 5 m/s. Another millisecond later, strong suction transports most of the bubbles back into the vicinity of the bone platelet, where they are strongly forced into the lateral regions. Some of the bubbles (mainly large ones) are mixed there through local turbulences near the bone. A few are again moved from the bone plate (perforation) into the lumen up to a maximal penetration depth of about 8 mm after 25 ms (Fig. 3).

In all laser systems, the high flow velocities shortly after the laser impulse in the fluid lead to rapid heat transport into the depth of the model, thus causing a rapid and brief temperature elevation. Subsequent convection promotes an intensive heat exchange, which results in equally rapid initial cooling and sometimes smaller secondary temperature rises due to secondary convection-dependent flow turbulences (Fig. 4). This is followed by a second phase of slower cooling mainly caused by heat conduction. Because of the intensive heat transport that is due to laser-induced convection flow in the fluid, the cooling process does not depend solely on the thermal properties of the cochlea model (bone, acrylic glass, and fluid). The duration of cooling is therefore contingent on the laser-specific maximal temperature and laser-specific convection as well as on the laser-unrelated system-dependent heat conduction.

Because it requires the highest energy, the Ho:YAG laser has the highest temperature rise and longest cooling phase to the baseline.

A footplate perforation can generally only be achieved with the pulsed lasers by multiple shots at the same application site, because only a small amount of tissue is ablated per application [13]. Figure 5a–c shows the temperature time courses of the pulsed laser systems examined and the respective pulse count needed for an adequate perforation. The first applications usually achieve only a small perforation or none at all; subsequent ones enlarge the perforation to a maximal diameter of 0.6 mm. This increases the fluid irradiation and thus the energy input in the fluid. Each laser impulse (except the nonperforating pulses) is followed by an increasing brief maximal temperature elevation from pulse to pulse accompa-

TABLE 1. Applied Laser Systems and Parameters

Laser system	Wavelength	Spot diameter	Pulse duration (full-width half-maximum)	Energy/impulse	Energy density	Number of pulses	Repetition rate	Total energy
Er:YAG	2,940 nm	600 µm	180 µs	70–150 mJ	25–51 J/cm^2	5	3 Hz	0.35–0.75 J
Er:YSGG	2,780 nm	550 µm	500 µs	70–115 mJ	30–48 J/cm^2	5	1 Hz	0.35–0.58 J
Ho:YAG	2,100 nm	550 µm	500 µs	200–330 mJ	85–140 J/cm^2	10	1 Hz	2–3.3 J
Excimer	308 nm	580 µm	120 ns	13 mJ	5 J/cm^2	500	10 Hz	6.5 J

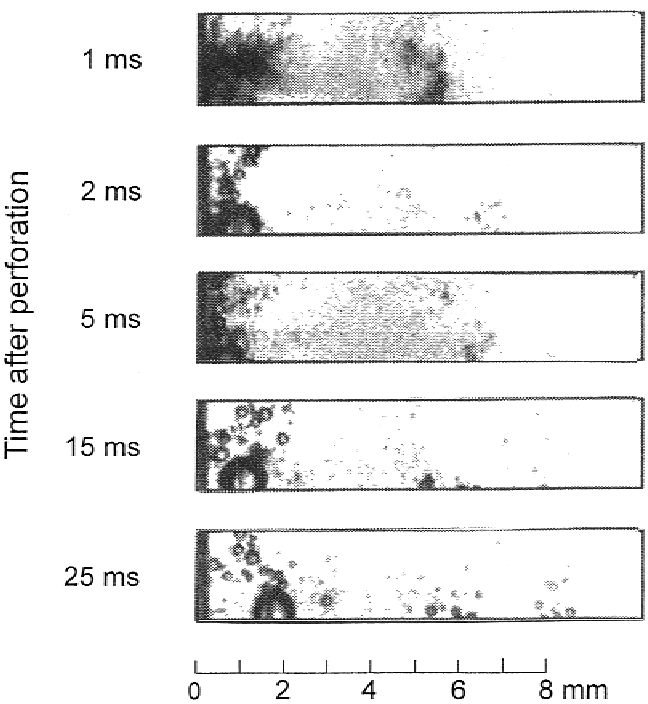

Fig. 3. High-speed videocamera (1,000 images/s) documentation of the Er:YSGG-laser-evoked heat transport mechanisms in the cochlea model based on induced gas bubbles. The laser beam hits the fluid from the left through the already-perforated bone platelet.

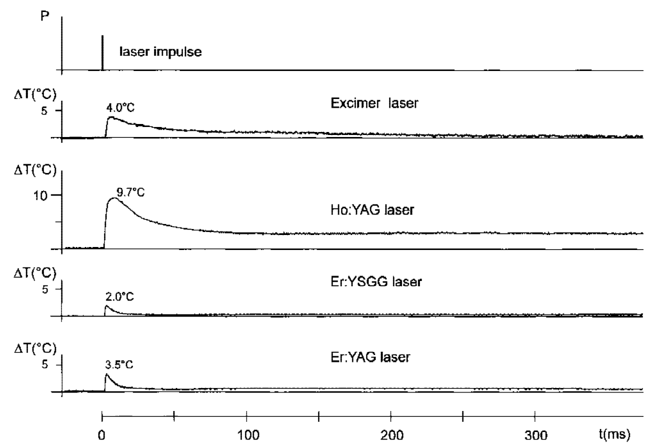

Fig. 4. Time course ΔT (°C) of temperature elevation in the fluid of the cochlea model at a distance of 2 mm behind the perforation with one laser impulse for the excimer laser (Q = 13 mJ, t = 120 ns (FWHM), H = 5 J/cm^2), Ho:YAG laser (Q = 250 mJ, t = 500 µs (FWHM), H = 105 J/cm^2), Er:YSGG laser (Q = 100 mJ, t = 500 µs (FWHM), H = 42 J/cm^2), and Er:YAG laser (Q = 95 mJ, t = 180 µs (FWHM), H = 34 J/cm^2).

nied by a gradual basic temperature rise, suggesting increased turbulent mixture of the fluid.

The maximal temperature elevations produced by the Er:YSGG and Er:YAG at a distance of 1 mm and 2 mm are 19.3°C and 3.6°C, respectively, with the Er:YSGG laser and 9°C and 5.5°C, respectively, with the Er:YAG laser (Fig. 5b,c). The basic temperature increases show no appreciable difference at a distance of 2 mm. They are 5°C and 2°C at a depth of 1 mm and 2 mm, respectively, with the Er:YSGG laser and gradually subside in the course of a few seconds (Fig. 5b). With the Er:YAG laser, they are somewhat lower, i.e., 4.2°C at 1 mm and 1.8°C at 2 mm (Fig. 5c).

The temperature time course of the Ho:YAG laser, on the other hand, exhibits less cooling between the individual laser pulses, which leads to a considerable increase in the basic temperature with that laser system (Fig. 5a). The increases are 17.5°C and 14.6°C at a distance of 1 mm and 2 mm, respectively. Renewed heat input after each laser impulse results in a stepwise elevation of the peak temperatures, which are higher after the last impulse with the Ho:YAG laser than with any of the other pulsed lasers examined (ΔT = 23.7°C and ΔT = 26.1°C at a distance of 1 mm and 2 mm, respectively).

The applied excimer laser radiating in the ultraviolet range caused a slight temperature increment of 4.5°C and 4°C at a distance of 1 mm and 2 mm behind the perforation. The high pulse count required for an adequate perforation (500 impulses), however, leads to a basic temperature of ΔT = 6°C at a distance of 1 mm and 2 mm, which is high in relation to the low maximal temperature.

Using the Er:YAG laser as an example, Figure 6 demonstrates the spatial heat distribution in the fluid of the cochlea model after laser appli-

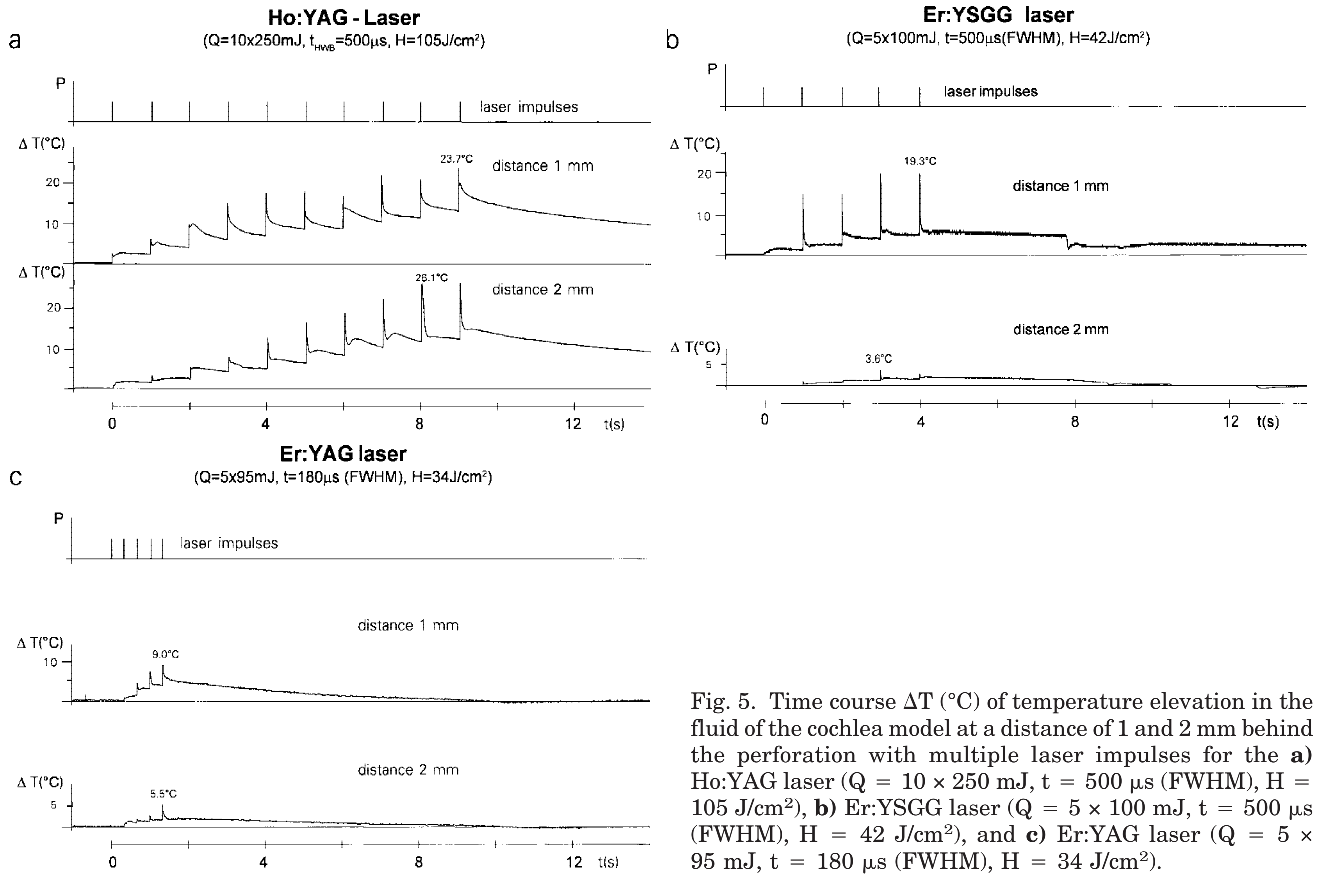

Fig. 5. Time course ΔT (°C) of temperature elevation in the fluid of the cochlea model at a distance of 1 and 2 mm behind the perforation with multiple laser impulses for the **a)** Ho:YAG laser (Q = 10 × 250 mJ, t = 500 µs (FWHM), H = 105 J/cm^2), **b)** Er:YSGG laser (Q = 5 × 100 mJ, t = 500 µs (FWHM), H = 42 J/cm^2), and **c)** Er:YAG laser (Q = 5 × 95 mJ, t = 180 µs (FWHM), H = 34 J/cm^2).

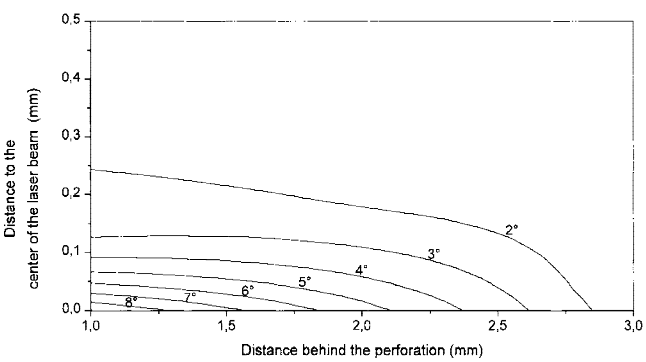

Fig. 6. Temperature field in the fluid of the cochlea model for perforating Er:YAG laser impulses (Q = 5 × 95 mJ, t = 180 µs (FWHM), H = 34 J/cm^2).

cation. The maximal temperatures occur in the immediate vicinity of the perforation, and the main spreading direction of heat transport is in linear continuation of the laser beam. The penetration depth of the heat, however, is very low on the whole.

With the Er:YAG laser, there is a nearly steady temperature decrease with distance at 2 mm and 3 mm from the perforation site in the examined energy-density range (30–50 J/cm^2), whereas nearly constant maximal temperatures result on application of the same energy densities with the Er:YSGG laser. The Ho:YAG laser also shows a temperature decrease with distance and strong scattering of the values at 1 mm and 2 mm behind the perforation (Fig. 7a). At a distance of 1 mm, high temperatures occurred at high energy densities (ΔT = 88°C at 125 J/cm^2, ΔT = 166°C at 140 J/cm^2), which are attributable to a direct irradiation of the temperature probe due to the lower absorption coefficient for water for the wavelength of 2.1 µm ($\alpha \geqslant 10$ cm^{-1}, ca. 400 times lower than for the wavelength of 2.94 µm (Er:YAG laser)) and thus higher penetration depth of Ho:YAG laser radiation in water. These data are not shown in the diagram.

In all systems examined, higher temperature increments result in connection with an increase in energy density, the degree of dependence differing markedly (Fig. 8). With the Ho:YAG laser, the higher energy density and pulse count required for an adequate perforation affect the temperature elevations, which are maximally 42.1°C. The temperature maxima that are lowest and only slightly dependent on the en-

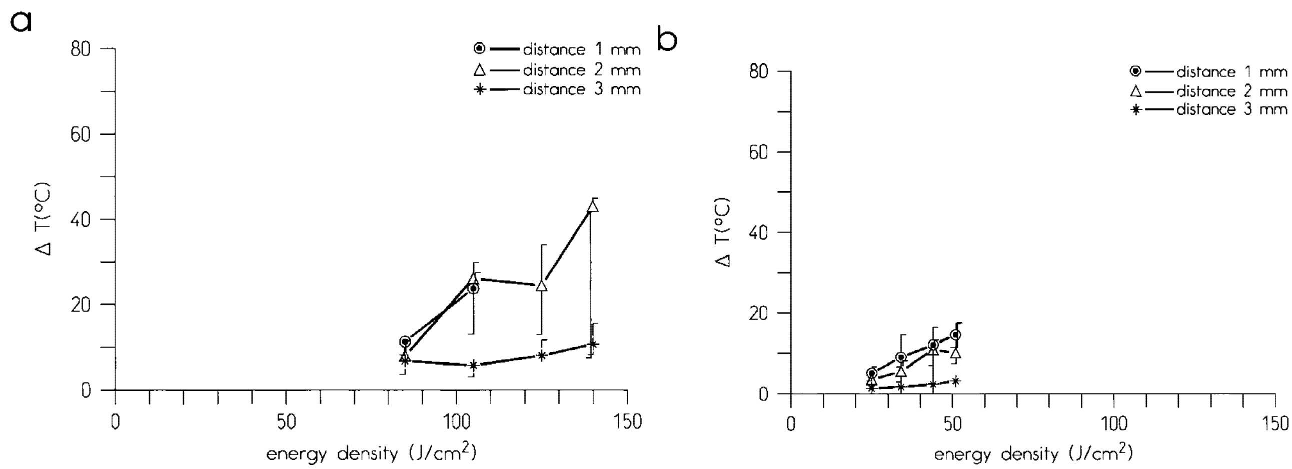

Fig. 7. Comparison of temperature elevations in the fluid of the cochlear model in relation to the power density and distance behind the perforation with the **(a)** Ho:YAG laser and **(b)** Er:YAG laser (median and range of 5 measured values).

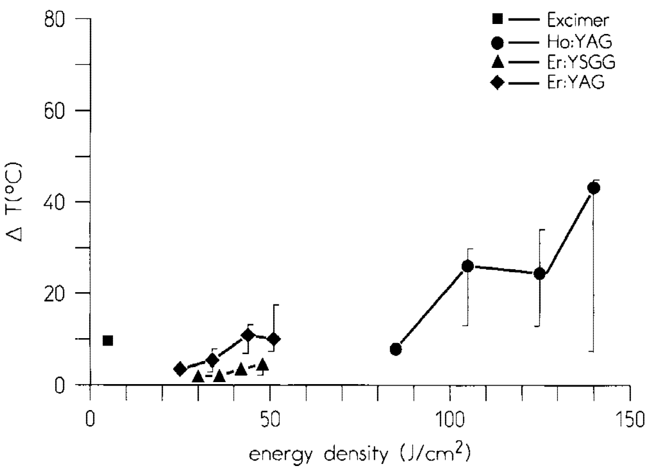

Fig. 8. Comparison of temperature elevations in the fluid of the cochlea model in relation to the energy density at a distance of 2 mm behind the perforation using various pulsed lasers with the different number of pulses necessary for sufficient perforation in each case (excimer: 500 pulses, Ho:YAG: 10 pulses, Er:YSGG: 5 pulses, Er:YAG: 5 pulses; median and range of 5 measured values).

ergy density result with the Er:YAG (ΔT = 10.9°C) and Er:YSGG (ΔT = 4.6°C) laser. The applied excimer laser, which could only be examined at one energy density, showed maximal temperature elevations of 10.3°C because of high total energies through a high pulse count.

DISCUSSION

Like the continuous-wave lasers, pulsed photoablative laser systems cause strong warming of the perilymph near the footplate by bundling the energy delivered at the surface of the fluid. The low single-pulse energy applied (≤0.3 J) causes initial temperatures at the application surface to be lower than with continuous-wave lasers. The short impulse duration of the laser radiation (120 ns to 500 μs) triggers an intensive heat transport, mainly in the form of convection flow, which leads to short-term and laser-system-specific temperature increases at greater distances from the application site. In addition, the strong flows result in an intensive mixture or heat distribution, which in turn causes rapid local cooling.

The short-term temperature elevations (above 100°C) directly at the perforation site behind the footplate are not directly measurable for lack of sufficiently small and rapid temperature sensors. In addition, there is the danger of direct irradiation of the sensor for measurements within the penetration depth of the laser beam, which leads to temperatures solely dependent on the absorption properties of the sensor and thus to incorrect measurements. For measurements at anatomically more relevant distances, e.g., 1 mm behind the perforation (beginning of membranous labyrinth components—sacculus, utriculus, and ductus cochlearis), direct irradiation of the sensor must be excluded [11]. In our experiments this phenomenon was observed only when applying the Ho:YAG laser at high energy densities (125–140 J/cm²).

The model was essentially approximated to cochlear conditions in its caloric parameters. Applicability of the results to the inner ear is ensured with respect to the heat transport mechanisms and the convection-determined maximal

temperatures for the space directly behind the footplate up to the sacculus and utriculus. By preventing linear spread, the cochlear convolutions, however, lead to deflections and mixing processes in the flow and thus to reduced heat penetration into the scala vestibuli. Simulation of heat transmission to the partially perfused tissue structures as found in situ is also inadequate but should only negligibly alter the caloric balance.

When water surfaces are irradiated by pulsed photoablative laser systems in the nanosecond range, plasma formation briefly gives rise to bubbles measuring up to 2 mm in diameter that implode again immediately after termination of the laser impulse (cavitation) [24,25]. Such high energy densities and short pulse half-widths were not applied with the lasers we used. This implied that plasma formation was negligible. Nevertheless, cavitation was possible. With our temporal kinematic resolution, we were not able to document such rapid processes (<100 μs). The bubbles we observed with a kinematic resolution of 1,000 images/s are generated mainly by air drawn in during the perforation and remain nearly stable during cooling. In contrast to the CO_2 laser in cw and superpulse mode [11], a distinctly higher number of visible bubbles form with pulsed lasers; they are also somewhat larger and spread in a more rapid and disordered manner in the cochlea model. This is an indirect indication of stronger convection than with continuous-wave lasers. On the one hand, the moving bubbles (flow velocity up to 5 m/s) may cause mechanical damage to the inner ear structure. This will be discussed in subsequent studies. On the other hand, the bubbles could increase the penetration depth of laser radiation into the vestibule and possibly damage inner ear structures by direct exposure. For instance, fluences higher than 80 J/cm^2 with the Er:YSGG laser are sufficient to penetrate a liquid layer of about 2 mm [19]. With the erbium laser, however, these high energy densities are not relevant for stapedotomy.

The heat delivered depends mainly on the total energy applied. With all systems, higher energy densities lead to higher temperatures. Besides the energy density, the pulse count and repetition rate also play a role in heating the fluid.

Whereas an adequate power density enables cw lasers to already achieve perforation at their first application, pulsed lasers cause only limited bone ablation with single impulses and thus do not usually achieve perforation at the first shot.

Only multiple application of laser radiation at the same site leads to the desired perforation. Thus, the first nonperforating applications of a pulse series do not irradiate the fluid directly, which results in less heating. Direct irradiation of the fluid in the cochlea model after perforation then leads to more marked temperature elevations with stronger scattering of maxima.

Through a summation effect, increasing the number of pulses raises the basic temperature and thus also the maximal temperatures. A higher repetition rate has the same effect, since there is less time for cooling between the pulses, and the initial temperature is not regained by the next application. Although the lowest system-specific repetition rates were used in our experiments, the basic temperature was increased to varying degrees even at the lowest settings of 1 Hz. The strongest temperature elevations are found with the Ho:YAG laser, because of slower cooling.

The differing temperature increases for the Er:YSGG and Er:YAG laser, given the same energy density per pulse and nearly the same absorption of the wavelengths in water, must be attributed to the influence of the system-specific pulse half-width (Er:YSGG: t = 500 μs (FWHM); Er:YAG: t = 180 μs (FWHM)). The approximately 3 times higher peak pulse power resulting from the shorter pulse half-width of the Er:YAG laser could have led to higher maximal temperatures through more intensive heat-exchange processes caused by stronger convection.

Apart from the temperature level reached, the time of tissue exposure to the temperature is important for the noxious potency of the thermic effect of laser irradiation on biological structures.

Several in vivo studies may be cited to answer the question of what magnitude of temperature increases in the cochlea (perilymph) leads to clinically relevant problems. Thus, Barnett [1] performed an in vivo study in cats and guinea pigs to examine the change of cochlear microphonic potentials (CM) in ultrasonic irradiation of the cochlea through the round window. Increasing the intracochlear fluid temperature to 5°C above normal body temperature had no effect on the microphonics, whereas a 7.5°C elevation reduced CM by approximately 75% after 2½ min. Raising the temperature by 10°C produced an 87% CM reduction within 1½ min. The series of experiments demonstrated that the CM impairment threshold was around 7°C above normal and that applying temperatures just above this threshold for short periods of 30 s or less caused a completely revers-

TABLE 2. Effective Laser Parameters for a 500 to 600-μm Stapes Footplate Perforation and Results and Temperature Elevations in the Perilymph at a Distance of 2 mm Behind the Perforation

Laser system	Beam diameter	Energy density	Rep. rate	Number of pulses	Application duration	Total energy	Maximal temperature elevation ΔT (median, maximum)	Basic temperature ΔT (median)	Heat exposure time
Er:YAG 2.94 μm, 180 μs	≈600 μm	34 J/cm^2	3 Hz	≈5	2 s	≈0.487	5.5°C (5.9°C)	1.8°C	≈10 s
Er:YSGG 2.78 μm, 500 μs	≈550 μm	42 J/cm^2	1 Hz	≈5	5 s	≈0.57	3.6°C (3.6°C)	1.8°C	≈10 s
Ho:YAG 2.1 μm; 500 μs	≈550 μm	105 J/cm^2	1 Hz	≈10	10 s	≈2.5 J	26.1°C (29.9°C)	14.6°C	≈30 s
Excimer 308 nm; 120 ns	≈580 μm	5 J/cm^2	10 Hz	≈200	20 s	≈2.6 J	9.7°C (10.3°C)	6.0°C	≈60 s

ible CM reduction of 15–20%. However, application of a suprathreshold temperature for more than 1–2 min led to a permanent CM depression. At temperature elevations of 7.5°C and 10°C, CM was reduced in response to all frequencies [1]. The temperature studies of Drettner et al. [3] in patients submitted to ultrasonic irradiation of the labyrinth for Meniere's disease (the ultrasonic transducer was directed toward the enchondral bone in the angle between the lateral and superior vertical semicircular canals) showed that temperature increases in the vestibule of up to 4°C for several minutes (about 7–8 min) were not associated with any deterioration of hearing in the majority of cases.

Despite differing heat vulnerability of biological tissue, the data of Moritz and Henrique [6] on the influence of temperature and duration on irreversible tissue damages are also an important orientational parameter for evaluating our results. Thus, brief heating (1 s) at 70°C (ΔT = 33°C) leads to the same tissue destruction as heating for 10 s at 58°C (ΔT = 21°C).

In our experiments a single perforating laser application led to in part only slight temperature elevations at a distance of 2 mm behind the perforation with all laser systems. The highest values are measured with the Ho:YAG (9.7°C) laser, the lowest with the Er:YSGG laser (2°C). These short-term maximal temperature elevations (<40 ms) are probably not sufficient to cause heat damage to tissue. The effect of the basic temperatures (ΔT < 3°C) that slowly subside over several seconds is likewise negligible as could be shown by Noyes et al. [16] in an animal study which confirmed that cochlear temperature elevation in the range of 3°C to 4.2°C did not impair rabbit outer hair cell function measured through distortion-product otoacoustic emissions at 1,806 to 8,691 Hz.

With multiple application, the highest temperature increments in the fluid of the cochlea model at a distance of 2 mm behind the perforation were likewise measured with the Ho:YAG laser and, in the effective laser range, were 29.9°C (median 26.1°C) above the initial temperature of 37°C (Table 2). Despite short thermic exposure times, inner-ear irritations cannot be excluded at these high temperatures (67°C). With the Er:YSGG and Er:YAG laser, on the other hand, the maximal temperature increases in the effective energy-density range are 3.6°C (median 3.6°C) and 5.9°C (median 5.5°C), respectively, and thus appear safe for the inner ear in view of the short heat exposure time (Table 2). The low temperature elevations of 10.3°C (median 9.7°C) with the excimer laser, on the other hand, can cause inner ear irritations through the long exposure time (about 10 s) due to the 500-fold application of the laser radiation with a high repetition rate of 10 Hz and a long application time (50s) (Table 2).

The slowly subsiding basic temperature associated with multiple application reaches maximal values of ΔT = 1.8°C at a distance of 2 mm behind the perforation with an exposure time of about 10 s for both the Er:YSGG and Er:YAG laser and can thus be excluded as a noxious factor (Table 2). Application of the Ho:YAG laser, on the other hand, leads to maximal basic temperatures of ΔT = 14.6°C, which subside in about 30 s (Table 2). The excimer laser exhibits basic temperature elevations of maximally 6°C over a period of about 60 s (Table 2). Inner-ear irritations are probable here due to the long heat exposure time, and irreversible inner-ear damage cannot be ruled out in the case of the Ho:YAG laser.

Conclusive clarification will only be provided by in vivo experiments. Our earlier experimental studies with the basal convolution of the guinea

TABLE 3. Characteristics Determining Laser Suitability in Stapedotomy With the Effective Parameters of Table 2*

Laser system	Ability to create an adequate perforation[a]	Low heat production in the perilymph	Inner-ear safety based on electrophysiological animal experiments[b]
Er:YAG 2.94 μm, 180 μs	+++	++	No data available
Er:YSGG 2.78 μm, 500 μs	+++	+++	+++
Ho:YAG 2.1 μm, 500 μs	++	–	–
Excimer 308 nm; 120 ns	–	++	–

*+++, very good; ++, good; +, satisfactory; –, poor.
[a]According to Jovanovic et al, 1997 [13].
[b]According to Jovanovic et al., 1995a–c [9–11].

pig cochlea as the site of application have documented the good tolerance and application safety of the pulsed Er:YSGG laser (Table 3) [10]. On the other hand, our results show that application of Ho:YAG laser radiation to the guinea pig cochlea with the laser parameters effective for footplate perforation acts unpredictably on the measured compound action potentials (CAP) [9]. These parameters already cause a slight to extreme irreversible CAP alteration in over 40% of the animals (Table 3). For the excimer laser as well, these investigations demonstrate that, despite low temperature elevations, the high total energies and very high power peaks needed for a perforation with this pulsed laser operating in the nanosecond range apparently result in extreme irreversible damage extending to hearing loss in the animals (Table 3) [9]. Beside heat as a noxious factor for inner ear structures with pulsed laser systems, damage could also be caused by the pressure phenomena occurring in the cochlea owing to high peak pulse powers. A final evaluation of our animal experiments can be made after the histological examination.

The literature contains only a few studies involving temperature measurements during the application of pulsed laser systems for stapedotomy [5,7,14,18,23].

Kautzky et al. [14] restricted themselves to measurements of temperature increases in the human stapes footplate at a distance of 0.3 mm to 0.5 mm from the application site of Ho:YAG laser radiation, which yielded a maximal value of 15°C at an energy density of 320 J/cm^2. There is no evidence supporting the assumption that heat conduction (lower in water than in bone) alone is important for the temperature increases in the perilymph and results in slight thermic loading on inner-ear structures. Convection in fluids makes possible a considerably more efficient and more rapid heat transport than heat conduction, so that it is of greater importance for the temperatures developing in the perilymph after laser irradiation, as also demonstrated by our results.

Hommerich and Hessel [7] likewise carried out temperature measurements in the cochlea model after application of the Ho:YAG laser. With a pulse energy of 250 mJ and a pulse half-width of 250 μs, there were temperature elevations of 10°C and <1°C at a distance of 0.2 mm and 1.5 mm behind the footplate after 1 to 2 laser applications.

In isolated human petrous bones and in vivo (chinchillas), Stubig et al. [23] investigated the temperatures found while irradiating the cochlea with a Ho:YAG laser coupled to a 200-μm core diameter fiber held less than 1 mm above the stapes surface, applying single-pulse energies of 30 mJ to 70 mJ with a 2-Hz pulse repetition rate and a pulse width of 250 μs. A temporal resolution of 3 measurements per second yielded temperature increases of maximally 9.7°C in human petrous bones at a distance of 1 mm to 1.5 mm from the perforation on application of up to 52 impulses with an energy of 65 mJ (total energy 3.4 J). Only slightly higher temperatures were measured during direct irradiation of the vestibule (ΔT = 12.2°C for 30 impulses at 55 mJ (175 J/cm^2) each). The in vivo measurements in the round window (90°, 2–2.5 mm away from the irradiation site) yielded maximally 6°C for a footplate perforation 300 μm in diameter with 4 to 15 single impulses at 40 mJ to 60 mJ (127–190 J/cm^2). From a thermic point of view, the Ho:YAG laser is therefore regarded as suitable for stapedotomy. However, Stubig et al. created smaller, clinically irrelevant

perforations with a diameter of 300 μm instead of 500 μm to 600 μm by applying lower energies with nearly the same energy densities we identified as effective for stapedotomy and performed their investigations with temperature sensors that were markedly slower than the ones we used in our study, which might explain the lower temperatures. Moreover, a lower temperature elevation in the vestibule is likewise simulated by not positioning the sensor in the middle of the convection flow.

Pfalz et al. [18] only performed temperature measurements at the surface of isolated bovine auditory ossicles after Er:YAG laser irradiation (pulse half-width 250 μs, beam diameter 200 μm) using a thermocamera. After application of a single 50-mJ impulse, they detected maximal temperature elevations of 48°C in the "crater center" that subsided to the initial value after 600 ms. They conclude from this that the erbium laser is suitable for footplate perforation from the thermic point of view.

Romano et al. [20] also measured temperature distributions on the bone surfaces (0.11-mm-thick cortical bone samples of human femur) during Er:YSGG laser radiation (pulse half-width 200 μs, beam diameter ca. 400 μm). They found temperature maxima of about 350°C at the ablation site. Histological evaluation of laser-treated footplates revealed a thermically damaged zone of <10 μm [19]. Temperature measurements performed in the inner ear model by the same group showed that ablation following Er:YAG laser treatment (pulse half-width 200 μs, beam diameter ca. 400 μm) does not lead to appreciable heating in the artificial perilymph [5]. With eight pulses at an energy density of 10 J/cm^2, maximal temperature increments of about 5°C were measured in the fluid directly behind the stapes footplate but outside the laser beam axis. A second sensor positioned at a distance of 3 mm behind the stapes footplate, also outside the laser beam axis, showed no temperature changes in the fluid. The lower temperatures are caused by the laterally positioned sensor not recording the maximal temperature range in the laser beam axis and by the lower energy densities used. They therefore conclude that the risk of thermal inner-ear damage due to erbium laser stapedotomy is negligible.

With regard to possible inner-ear damage by thermic loading during laser stapedotomy, it is harmless from a thermic point of view to apply the Er:YSGG laser for laser stapedotomy in a relatively broad energy density range. Laser systems with short impulse durations lead to stronger heating at the same energy density because of higher peak pulse powers. This holds true for the examined Er:YAG laser. Although the Er:YAG laser is suitable for stapedotomy from a thermic point of view, the suitability of the Ho:YAG laser is doubtful because of the higher energy density and pulse rate required for sufficient perforation and the resultant higher and longer-lasting temperature increases in the perilymph. Furthermore, direct irradiation of cochlear structures cannot be excluded.

Although perforations with the applied excimer laser lead to slight temperature elevations, this system does not appear suitable for stapedotomy because of the long heat exposure time (60 s) and the low ablation rates at the stapes.

Considering their high ablation rates at the stapes and only slight thermic side effects, the pulsed erbium lasers could prove to be a useful alternative to the continuous-wave thermic CO_2 lasers.

For a definitive statement on the most effective and safest laser system for stapes surgery, it is of the utmost importance to also investigate the acoustic stress (pressure effects in the perilymph) during laser stapedotomy. This will be examined in subsequent studies.

REFERENCES

1. Barnett SB. The influence of ultrasound and temperature on the cochlear microphonic response following a round window irradiation. Acta Otolaryngol 1980; 90:32–39.
2. Biyikli S, Modest M, Tarr R. Measurements of thermal properties for human femora. J Biomed Mat Res 1986; 20:1335–1345.
3. Drettner B, Johnson S, Sjöberg A, Stahle J. Some thermal effects of ultrasound on the inner ear. Acta Otolaryngol 1967; 64:464–476.
4. Fischer R, Schönfeld U, Jovanovic S, Jaeckel P. Thermische Belastung des Innenohres durch verschiedene Lasertypen bei der Laser-Stapedotomie. Eur Arch Otolaryngol 1992; Suppl II:251–253.
5. Frenz M, Romano V, Pratisto H, Weber HP, Altermatt HJ, Felix D, Grossenbacher R. Stapedotomy: new results of the erbium laser. In: ORL "Aktuelle Probleme der Otolaryngologie." Bern: Hans Huber Verlag, 1994, pp 335–343.
6. Helfmann J, Brodzinski T. Thermische Wirkungen. In: Berlien HP, Müller G (Hrsg). Angewandte Lasermedizin II-3.3. Ecomed Verlag Landsberg, 1992, pp 1–8.
7. Hommerich C, Hessel S. Untersuchungen mit dem Holmium: YAG Laser an Amboß und Steigbügel. Eur Arch Otolaryngol 1991; (Suppl)II:280.

8. Jovanovic S, Schönfeld U, Prapavat V, Berghaus A, Fischer R, Scherer H, Müller G. CO_2 laser in stapes surgery. Proc SPIE 1993; 1876:17–27.
9. Jovanovic S, Anft D, Schönfeld U, Prapavat V, Berghaus A, Scherer H, Müller G. Suitability of pulsed laser systems for stapedotomy—an animal experimental study. Lasermedizin 1995; 11:140–149.
10. Jovanovic S, Anft D, Schönfeld U, Berghaus A, Scherer H. Experimental studies on the suitability of the erbium laser for stapedotomy in an animal model. Eur Arch Otorhinolaryngol 1995; 252:38–39.
11. Jovanovic S, Schönfeld U, Fischer R, Döring M, Prapavat V, Müller G, Scherer H. Thermische Belastung des Innenohres bei der Laserstapedotomie Teil I: Kontinuierlich strahlende Laser. HNO 1995; 43:702–709.
12. Jovanovic S, Schönfeld U, Prapavat V, Berghaus A, Fischer R, Scherer H, Müller G. Effects of continuous-wave laser systems on stapes footplate. Lasers Surg Med 1996; 19:424–432.
13. Jovanovic S, Schönfeld U, Prapavat V, Berghaus A, Fischer R, Scherer H, Müller G. Effects of pulsed laser systems on stapes footplate. Lasers Surg Med 1997; 21: 341–350.
14. Kautzky M, Trödhan A, Susani M, Schenk P. Infrared laser stapedotomy. Eur Arch Otorhinolaryngol 1991; 248: 449.
15. Lesinski SG. Lasers for otosclerosis. Laryngoscope 1989; 99(Suppl 46):1–24.
16. Noyes WS, McCaffrey TV, Fabry DA, Robinette MS, Suman VJ. Effect of temperature elevation on rabbit cochlear function as measured by distortion-product otoacoustic emissions. Otolarngol Head Neck Surg 1996; 115:548–552.
17. Perkins RC. Laser stapedotomy for otosclerosis. Laryngoscope 1980; 90:228–241.
18. Pfalz R, Hibst R, Bald N. Suitability of different lasers for operations ranging from the tympanic membrane to the base of the stapes. Adv Otorhinolaryngol 1995; 49:87–94.
19. Pratisto H, Frenz M, Ith M, Romano V, Felix D, Grossenbacher R, Altermatt H, Weber H. Temperature and pressure effects during erbium laser stapedotomy. Lasers Surg Med 1996; 18:100–108.
20. Romano V, Rodriguez R, Altermatt H, Frenz M, Weber H. Bone microsurgery with IR-lasers: a comparative study of the thermal action at different wavelengths. SPIE 1993; 2077:87–97.
21. Segas J, Georgiadis A, Christodoulou P, Bizakis J, Helidonis E. Use of the excimer laser in stapes surgery and ossiculoplasty of middle ear ossicles: Preliminary report of an experimental approach. Laryngoscope 1991; 101: 186–191.
22. Silverstein H, Rosenberg S, Jones R. Small fenestra stapedotomies with and without KTP laser: A comparison. Laryngoscope 1989; 99:485–488.
23. Stubig IM, Reder PA, Facer GW, Rylander HG, Welch AJ. Holmium: YAG laser stapedotomy: preliminary evaluation. Proc SPIE 1993; 1876:10–19.
24. Tomita Y, Shima A. High-speed photographic observations of laser-induced cavitation bubbles in water. Acustica 1990; 71:161–171.
25. Vogel A, Hentschel W, Holzfuss J, Lauterborn W. Cavitation bubble dynamics and acoustic transient generation in ocular surgery with pulsed neodymium:YAG lasers. Ophthalmology 1986; 93:1257–1269.